

Reduction of Resonant RF Heating in Intravascular Catheters Using Coaxial Chokes

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The incorporation of RF coils into the tips of intravascular devices has been shown to enable the localization of catheters and guidewires under MR guidance. Furthermore, such coils can be used for endoluminal imaging. The long cable required to connect the coil with the scanner input inadvertently acts as a dipole antenna which picks up RF energy from the body coil during transmit. Currents are induced on the cable which can lead to localized heating of surrounding tissue. Cables of various lengths were measured to determine if a resonance in the heating as a function of cable length could be found. Coaxial chokes with a length of $\lambda/4$ were added to coaxial cables to reduce the amplitude of the currents induced on the cable shield. A 0.7-mm diameter triaxial cable, small enough to fit into a standard intravascular device, was developed and measured both with and without a coaxial choke. It is demonstrated that resonant heating does occur and that it can be significantly reduced by avoiding a resonant length of cable and by including coaxial chokes on the cable. *Magn Reson Med* 43:615–619, 2000. © 2000 Wiley-Liss, Inc.

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With the advent of MR scanners with open magnet designs and shorter magnet lengths, much interest has been expressed in performing vascular interventions using MR guidance in place of X-ray fluoroscopy. To provide reliable visualization of catheters and guidewires during vascular interventions under MR guidance, several techniques have been introduced which incorporate an RF antenna into the catheter or guidewire (1–4). In addition, an RF antenna in a catheter can be used for endoluminal imaging (5), which may prove useful for characterizing atherosclerotic changes (6). One of the significant concerns raised for such devices is the possibility of localized heating due to the antenna itself or the long coaxial cable connecting the antenna to the scanner.

Temperature measurements around intravascular devices demonstrate that the heating is not related to transitions in the gradient fields, but is instead due exclusively to coupling with the transmit RF field from the body coil (7). Measurements of actively visualized catheters and guidewires have shown significant temperature rises (up to 68°C under extreme conditions) near the catheter coil at 1.5 T when using RF-intense imaging sequences, such as fast spin echo (7,8).

If the body coil is used for transmit, part of the energy can be coupled into the catheter coil or into the cable

connecting the coil to the scanner, with the result that local tissue heating can occur around the coil or cable. The coupling to the coil itself can be reduced by detuning the coil during transmission, which can be achieved either passively with crossed diodes or actively with PIN diodes. If the decoupling fails, the increased specific absorption rate (SAR) can induce significant local heating around conventional, external surface coils. However, previous work has demonstrated that for the small-area coils used in interventional intravascular work, the primary heating mechanism is via coupling to the long cable between the coil and scanner and not due to magnetic field coupling to the coil itself (8). This was demonstrated by showing that the heating did not depend on whether the coil was tuned or not tuned and that the heating was not dependent on the orientation of the coil within the B_1 field. Also, long cables without any RF coil showed significant heating, whereas RF coils without any cable attached showed no heating. Maximal heating was obtained when the cable was along the bore wall, whereas a cable near isocenter showed much less heating, indicating coupling to the E_1 field of the body coil. In all cases, the only measurable heating occurred near the tip of the cable/coil combination. No significant heating was ever detected along the cable (7,8). This was presumably due to concentration of electric fields at the sharp tip of the cable/coil combination (9).

The coupling to the cable cannot be reduced by detuning the coil; however, several methods for reducing unwanted shield currents are available using RF chokes. In the case of a coaxial cable, these chokes present a high impedance to currents flowing on the outer surface of the shield, but present no impedance to the differential currents flowing in the cable itself between the inner conductor and the inner surface of the shield. One possibility is to wind a loop or loops in the cable. In some cases, a ferrite core is used to increase the inductance of the windings (10), which would not be practical in an MR scanner. The tuning capacitance of the choke can come from the self-capacitance of the windings (10), or from a discrete capacitor across the loops (11).

Several techniques have also been developed to translate between a balanced antenna and an unbalanced line, such as a coaxial cable. This is also achieved by creating a high impedance to currents on the outer surface of the shield. In this case the choke is often called a balun. One possibility is to use a coaxial sleeve with a length of $\lambda/4$ (12). Because of the extreme restrictions on the diameter of interventional devices, coaxial chokes with lengths close to $\lambda/4$ have been used in microwave hyperthermia (13–15) and have been proposed for MR intravascular loopless antennas (3) to help terminate one-half of the dipole antenna.

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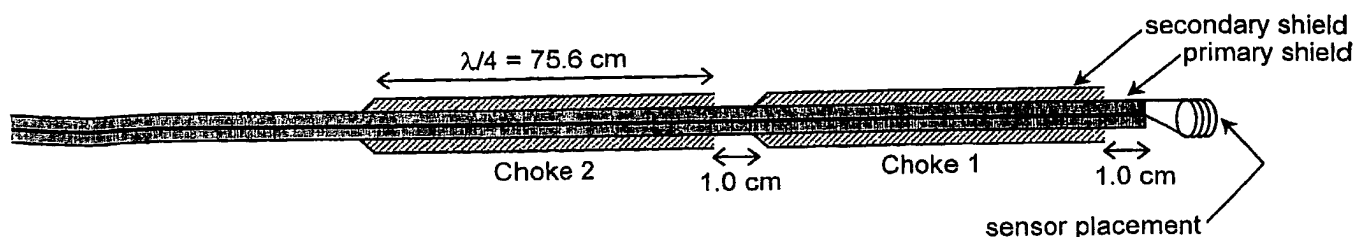


FIG. 1. Schematic showing implementation of two coaxial chokes in series using a Belden 9222 triaxial cable. Each choke is prepared by soldering a short between the primary and secondary shields of the triaxial cable at one end and removing the secondary shield at the other end. The space between the primary and secondary shields acts as a waveguide which translates the short into a high impedance at the open end of the choke.

The purpose of this investigation was to examine the effectiveness of coaxial chokes for eliminating the unwanted currents on the outside of the coaxial cable shield, since these currents lead to the localized tissue heating. Finally, a triaxial cable small enough to fit in a standard interventional device such as a 6 French catheter was designed and tested.

METHODS AND RESULTS

Temperature measurements were conducted on two separate sets of cables. The first set consisted of large-diameter triaxial cable. This set was used to test the effectiveness of various coaxial choke configurations. The second set was made up of small-diameter triaxial cable. This set was used to test the effectiveness of a single coaxial choke on a cable thin enough to fit into a standard intravascular device. In all cases a small solenoidal coil was soldered to the end of the cable under test.

Scanning was performed on a 1.5T Signa LX (GE Medical Systems, Milwaukee, WI). A Plexiglas phantom was constructed to support the cables and coils and allowed either full or partial immersion of the devices in physiologic saline solution (0.9% NaCl). A 2D fast-spin echo sequence using the body coil for transmit was used to induce RF heating near the coils. The parameters were the same as used in previous studies (7,8,16): axial sections, TR = 2000 ms, two echoes, TE1 = 16 ms, TE2 = 80 ms, echo train length = 8, FOV = 40 cm, slice thickness = 3 mm, slice spacing = 1.5 mm, number of slices = 13, matrix = 256 × 128, number of excitations = 4. The transmit gain was intentionally increased by 7.2 dB above the value selected by the scanner calibration software to obtain the maximum transmit gain available from the scanner, approximately 16 kW.

The phantom and cables were kept parallel to the main axis of the scanner bore. The cables were elevated to the isocenter of the magnet and moved laterally until they were as close to the bore wall as was possible (6.5 cm), as constrained by the size of the phantom. This position provides near maximal heating (8).

A fluoroptic thermometer (Model 790, Luxtron, Santa Clara, CA) was used to monitor the temperature near the RF coil. The temperature could be monitored in real time simultaneous with scanning, with four updates per second. The temperature sensor was placed at the distal opening of the small solenoidal coil soldered to the end of the

cable under test (Fig. 1). The microcoil facilitated repeatable placement of the fluoroptic temperature sensor (dia. 0.5 mm) at a position which had shown maximal heating in previous studies (7,8). Those previous studies had failed to show any measurable heating along the cable away from the tip, so measurements were only performed at the tip.

Large-Diameter Triaxial Cable

Coaxial chokes were made by soldering together the primary and secondary shields of a triaxial cable (9222, Belden Wire and Cable Co., Richmond, IN) and removing the secondary shield $\lambda/4$ away from the solder point (Fig. 1). A solenoidal coil with 10 windings and an outer diameter of 1 mm, similar to those used for MR tracking (1,17), was soldered onto the end of the 6 mm diameter triaxial cable between the inner conductor and the primary shield. To form the coil, copper wire with a diameter of 0.15 mm was used.

Three separate test cables were used to determine the appropriate choke length by using a vector impedance meter (Model 4193A, Hewlett Packard, Palo Alto, CA) to measure the impedance between the primary and secondary shields when looking into the open end of the choke. The choke length was chosen such that the phase of the impedance reached 0° and became purely real, indicating a length of $\lambda/4$ in the dielectric between the primary and secondary shields. The length was determined to be 75.6 ± 0.14 cm, indicating a relative permittivity of 2.4 at 64 MHz. The Belden catalog specifies the insulation as polyethylene, and the measured permittivity is very close to the nominal permittivity of 2.3 available in the literature (18). The mean impedance of the choke was $525 \pm 11 \Omega$.

For temperature measurements, four triaxial cables were prepared, each with a length of 250 cm. The first had no chokes and had only the microcoil attached. The second was identical to the first, but had the outer insulation stripped from the entire length of the cable. The third was fully insulated, but had a single choke with the opening of the choke placed 1 cm from the solder joint of the microcoil (Fig. 1). The fourth had two chokes in series, the first with the opening placed 1 cm from the solder joint of the microcoil, and the second with the opening placed 1 cm from the shorted end of the first choke (Fig. 1).

The microcoils and 5 cm of cable were immersed in 0.9% saline solution, while the rest of the cable was kept

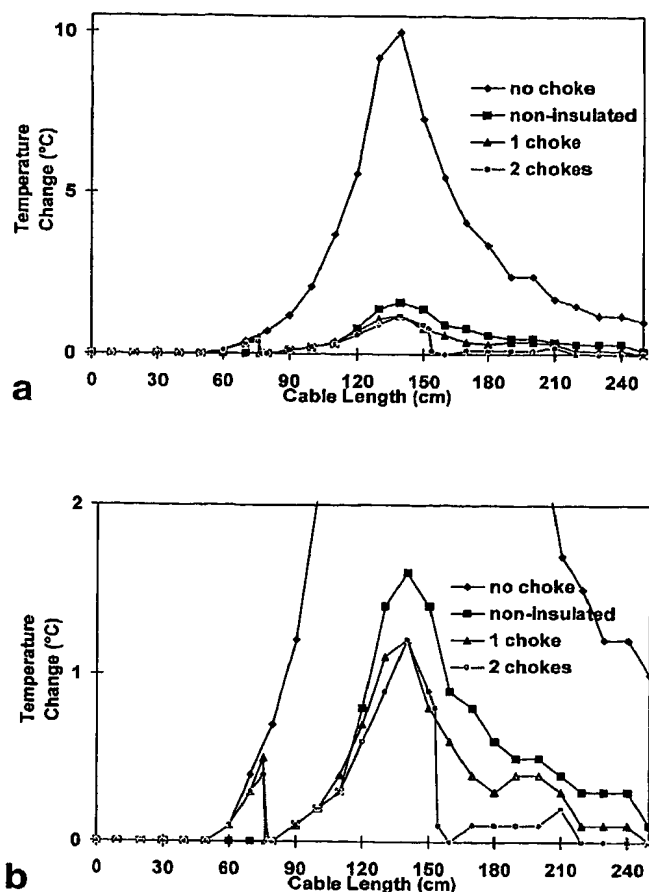


FIG. 2. a: Temperature increase as a function of cable length for various Belden 9222 triaxial cables. b: The same graph as in a, but with an expanded vertical scale. Note the excellent match between the cable with one choke and the cable with two chokes after the first of the two chokes is cut off at 154 cm. Note also the excellent match between the cable with no choke, the cable with one choke, and the cable with two chokes after all chokes are removed at 77 cm.

in air. The length of each cable was progressively shortened from its initial length of 250 cm in 10 cm steps and the maximum temperature increase at the solenoid was recorded for each length.

Figure 2 shows the measured temperature increase as a function of cable length for the four cables. The resonance was at 140 cm, which is approximately $0.3 \lambda_{\text{air}}$, where λ_{air} is the wavelength in air at 64 MHz. The discontinuity at 77 cm for the cable with one choke occurred as the shorted end of Choke 1 was cut off the end of the cable. The discontinuities at 154 cm and 77 cm for the cable with two chokes occurred as the shorted ends of Choke 2 and Choke 1 were cut off the end of the cable, respectively. The maximum temperature increase of the cable with two chokes before the first choke was cut off was 0.2°C.

Small-Diameter Triaxial Cable

Finally, a 0.7-mm diameter triaxial cable was designed and constructed (Fig. 3). Two test cables were used to deter-

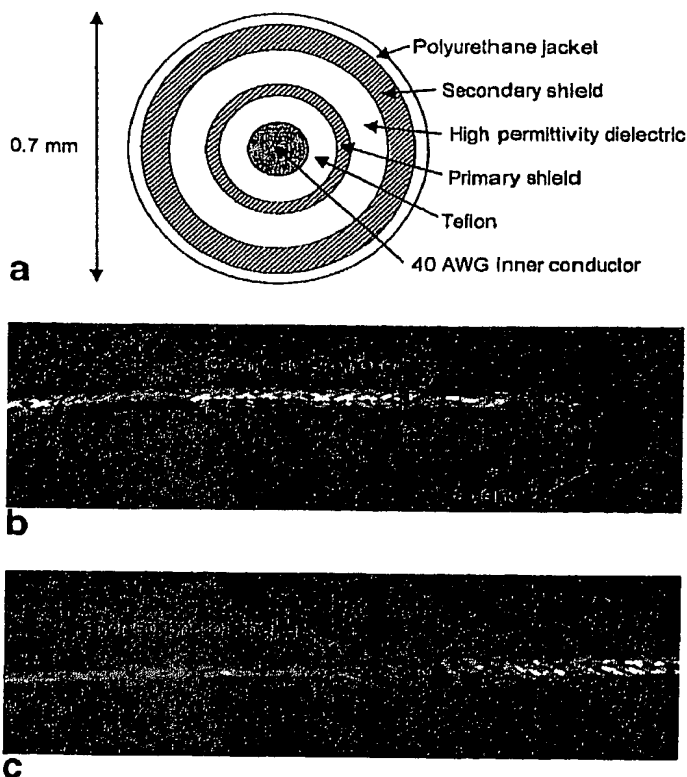


FIG. 3. a: Cross-section of a prototype triaxial cable with a diameter of 0.7 mm. b: Close-up of the coaxial choke opening and the solenoid coil at the end of the cable. c: Close-up of the laser-soldered short between the primary and secondary shields.

mine the appropriate choke length using the vector impedance meter. The $\lambda/4$ length was determined to be 55.8 ± 0.7 cm, indicating a relative permittivity of 4.4 at 64 MHz. The mean impedance of the choke was $188 \pm 1.4 \Omega$.

For temperature measurements, one cable was prepared with no choke. Two others were prepared with a single choke similar to Choke 1 in Fig. 1, but with the opening of the choke only 3 mm from the solder joint of the coil. A solenoidal coil with two layers of eight windings, for a total of 16 windings, was soldered between the inner conductor and the primary shield. In all cases the cables were tested in air and a resonant length (140 cm) was used to generate worst-case heating. Table 1 gives the temperature changes measured at the distal end of the microcoil for the three 0.7-mm diameter cables.

The temperature along the cable, both proximal and distal to the choke opening, was measured for one of the 0.7-mm diameter cables with a choke. Figure 4 shows the

Table 1
Maximum Temperature Increase Measured at the RF Coil of a 0.7 mm-Diameter Triaxial Cable

| Cable configuration | Temperature increase |
|---------------------|----------------------|
| Without choke | 55°C |
| With 1 choke | 3.7°C |
| With 1 choke | 3.4°C |

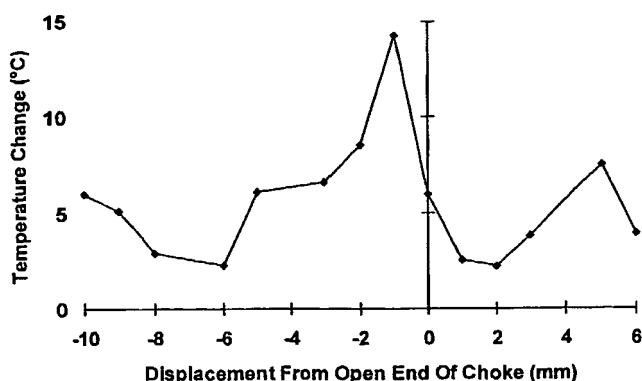


FIG. 4. Temperature increase near the opening of a choke on a 0.7-mm diameter triaxial cable. Positive displacements are in the direction of the solenoidal coil, and +6 mm corresponds to the distal tip of the solenoid where the maximum temperature increase is normally measured for cables without chokes.

temperature variation along the cable near the opening of the choke of the second cable in Table 1.

DISCUSSION

It has been demonstrated that localized heating near the tips of long, thin conducting objects placed in the MR scanner can be significantly reduced by a judicious choice of object length to avoid resonance, and by the incorporation of coaxial chokes to disrupt the standing wave pattern on the exterior conducting surface. Other safety concerns, including the possibility of nerve stimulation in the vicinity of intravascular catheters, were not investigated.

It is unclear what the appropriate safety limits are for the localized heating measured near intravascular RF coils. The United States Food and Drug Administration specifies a peak SAR of 8 W/kg in any 1 g of tissue (19). The National Radiological Protection Board in the United Kingdom specifies a 1°C temperature increase in any 1 g of tissue (20). However, the temperature increases for the types of coils measured here are generally limited to volumes less than 1 g of tissue (7). A reasonable goal appears to limit any temperature increase to less than 4°C, thereby avoiding any potential tissue damage, even with long-term exposure (21).

The wavelength of an insulated antenna in a conducting medium is a complicated function of insulation thickness and medium properties (22). It will, however, be shorter in saline solution than in air. During the course of an actual intervention, the fraction of catheter inserted into the patient and the fraction in air outside the patient will vary. As the boundary conditions are changed, so will the resonant length. Therefore, a prudent safety strategy would not rely solely on avoidance of a resonant heating length, since this length will depend on catheter insertion. The resonant length for the cable in air for these experiments was 140 cm, or $0.3 \lambda_{\text{air}}$, also an unexpected result, since resonance would be expected at $0.5 \lambda_{\text{air}}$. The reason for this discrepancy is unknown, but it may be due to the fact that 5 cm of cable was immersed in saline for performing temperature measurements in a conducting medium. Reflections at the

air/saline boundary may have shifted the resonant condition at the tip of the cable.

Figure 2 shows that one of the easiest ways to attenuate the standing wave on the outside of the cable is to remove the insulation and expose the wave to the lossy medium of NaCl solution or, in the case of in vivo interventions, blood. However, an uninsulated cable may be undesirable for manufacturing reasons or for safety concerns, such as electrical isolation from the patient.

As can be seen in Fig. 2 and Table 1, the large-diameter triaxial cable showed significantly less peak heating than the small-diameter cable under the conditions of resonance (a length of 140 cm in air). This effect was repeatedly found in other, similar, measurements. Smaller-diameter objects show higher peak heating than larger-diameter objects, probably because of the higher concentration of E-fields near thin, sharp objects (9), and possibly because of a reduced capacity to conduct heat away from the area via the conducting cable itself. Intravascular imaging coils which consist of a loop with a length of 40 mm and a width of 6 mm have also demonstrated much lower heating (16), probably due to the rounded end of the coil, which prevents a high concentration of E-fields.

Table 1 shows that the temperature reduction is repeatable when the choke is chosen to be the correct length. It also shows that even when the object has a resonant length, the temperature can be controlled through the use of coaxial chokes. A second choke added to the cable would be expected to reduce the maximum heating even further, as it did for the large-diameter Belden cable.

When a choke is incorporated into the cable, the point of maximum E-field amplitude is displaced to near the opening of the choke (Fig. 4), where the high impedance of the choke significantly reduces the current in the standing wave at that point on the cable, but simultaneously increases the E-field. However, the heating is not as localized as for the cable with no choke (7), so the temperature increases are spatially distributed and the peak temperature increase is reduced. In an actual catheter, the choke openings will be further isolated from the patient by the dielectric of the catheter itself. For cables without chokes, insulating the tip of the coil is not an option, since a guidewire lumen must be maintained. Further experiments will be required to quantify the heating that occurs around a cable inside an actual catheter. The temperature was measured only near the tip of the cable for these experiments, not along its entire length. A catheter intended for human use would also need to be tested over its entire length under a variety of experimental conditions to verify that there are no other hot spots.

To meet the unique size and geometry constraints of interventional catheters and guidewires, the coaxial chokes used in this work seem especially well suited when compared to alternatives. It is possible to create chokes with loops in the cable or by using discrete passive components to create blocking circuits. However, loops would not be compatible with interventional catheters, and discrete circuits would significantly add to the minimum diameter of the device. The extra layers of dielectric and shield added to the 0.7-mm diameter cable increased the cable diameter by only 0.3 mm, implying a 1 French in-

crease in the minimum diameter of the interventional device.

Another advantage of the coaxial choke is that its length is determined by the dielectric between the primary and secondary shields and is not dependent on the medium in which the cable is immersed. Therefore, its effectiveness is not dependent on the insertion depth of the catheter into the body. The dielectric between the primary and secondary shield of the 0.7-mm diameter cable used in these experiments was specifically chosen to obtain as high of permittivity as possible. The intent was to reduce the length of the choke. The experiments showed that a relative permittivity of 4.4 was obtained, which reduced the choke length to 56 cm. A short choke length is desirable to incorporate as many chokes as possible onto a given length of cable. Each choke adds additional impedance to surface currents and reduces the standing wave amplitude. Capacitors could be added to the coaxial chokes to reduce their length (23), but discrete components do not seem practical for devices with such small diameters.

The cable design shown here is well suited for intravascular RF antennas which are fairly short, so that the opening of Choke 1 is close to the distal end of the cable/coil combination. It is unknown how effective such a choke would be when used in combination with resonant dipole antennas (3), where the length of the antenna itself might approach the resonant length in saline solution.

Although the primary focus of this work was on intravascular catheter coils, the results showing resonant heating and the strategies to reduce it would also be applicable to other long conducting objects brought into patient contact, such as EKG and pacemaker leads, endoscopes, long needles, etc. Similar RF heating effects have been shown with EKG cables (24), as well as surface coil leads (25). Cable chokes are used routinely on surface coil cables as part of patient safety. Some EKG cables use high-impedance conductors, which do not appear to be an option for intravascular catheter imaging because of the extremely small signal amplitude.

The temperature increases presented here were measured under extreme conditions, which were selected to enhance the heating as much as possible, including increasing the scanner transmit gain to its maximum value, roughly 16 kW. By avoiding resonant lengths and incorporating multiple coaxial chokes, it is unlikely that unacceptable heating will occur under normal scanning conditions. However, each intravascular catheter design should be tested in its final configuration. These strategies for reducing heating to an acceptable level are an important step toward performing vascular interventions safely under MR guidance at higher field strengths.

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REFERENCES

1. Dumoulin CL, Souza SP, Darrow RD. Real-time position monitoring of invasive devices using magnetic resonance. *Magn Reson Med* 1993;29:411-415.
2. McKinnon GC, Debatin JF, Leung DA, Wildermuth S, Holtz DJ, von Schulthess GK. Towards active guidewire visualization in interventional magnetic resonance imaging. *MAGMA* 1998;4:13-18.
3. Ocali O, Atalar E. Intravascular magnetic resonance imaging using a loopless catheter antenna. *Magn Reson Med* 1997;37:112-118.
4. Ladd ME, Zimmermann GG, Quick HH, Debatin JF, Boesiger P, von Schulthess GK, McKinnon GC. Active MR visualization of a vascular guidewire in vivo. *J Magn Reson Imaging* 1998;8:220-225.
5. Atalar E, Bottomley PA, Ocali O, Corrolo LC, Kolonen MD, Lima JA, Zorhouni EA. High resolution intravascular MRI and MRS by using a catheter receiver coil. *Magn Reson Med* 1996;36:596-605.
6. Zimmermann GG, Erhart P, Schneider J, von Schulthess GK, Schmidt M, Debatin JF. Intravascular MR imaging of atherosclerotic plaque: ex vivo analysis of human femoral arteries with histologic correlation. *Radiology* 1997;204:769-774.
7. Wildermuth S, Dumoulin CL, Pfammatter T, Maier SE, Hofmann E, Debatin JF. MR-guided percutaneous angioplasty: assessment of tracking safety, catheter handling and functionality. *Cardiovasc Intervent Radiol* 1998;21:404-410.
8. Ladd ME, Quick HH, Boesiger P, McKinnon GC. RF heating of actively visualized catheters and guidewires. In: *Proc ISMRM, 6th Scientific Meeting and Exhibition*, Sydney, 1998. p 473.
9. Ramo S, Whinnery JR, van Duzer T. *Fields and waves in communications electronics*. New York: John Wiley & Sons; 1984. p 817.
10. Harper ET. Progress in broadband choke design. *Proceedings of the 21st electronic components conference*. New York: IEEE; 1971. p 59-66.
11. Edelstein WA, Iben IET, Mueller OM, Uzgrlis EE, Philipp HR, Roemer PB. Radiofrequency ground heating for soil remediation: science and engineering. *Environ Prog* 1994;13:247-252.
12. Sinnema W. *Electronic transmission technology*. Englewood Cliffs: Prentice Hall; 1988. p 380.
13. Hurter W, Reinhold F, Lorenz WJ. A dipole antenna for interstitial microwave hyperthermia. *IEEE Trans Microw Theory* 1991;39:1048-1054.
14. Wong TZ, Tremblay BS. A theoretical model for input impedance of interstitial microwave antennas with chokes. *Int J Radiat Oncol Biol Phys* 1994;28:673-682.
15. Lin JC, Yu Jin W. The cap-choke catheter antenna for microwave ablation treatment. *IEEE Trans Biomed Eng* 1996;43:657-660.
16. Quick HH, Ladd ME, von Schulthess GK, Debatin JF. Heating effects of an intravascular imaging catheter. *Brussels: MAGMA*; 1997. p 187.
17. Leung DA, Debatin JF, Wildermuth S, McKinnon GC, Holtz D, Dumoulin CL, Darrow RD, Hofmann E, von Schulthess GK. Intravascular MR tracking catheter: preliminary experimental evaluation. *AJR Am J Roentgenol* 1995;164:1265-1270.
18. Marshall SV, Skitek GG. *Electromagnetic concepts and applications*. Englewood Cliffs, NJ: Prentice Hall; 1987. p 505.
19. Athey TW. FDA regulation of the safety of MR devices: past, present, and future. *Magn Reson Imaging Clin N Am* 1998;6:791-795.
20. National Radiological Protection Board. Revised guidance on acceptable limits of exposure during nuclear magnetic resonance clinical imaging. *Br J Radiol* 1983;56:974-977.
21. Chou CK. Radiofrequency hyperthermia in cancer therapy. In: Bronzino JD, editor. *The biomedical engineering handbook*. Boca Raton, FL: CRC Press; 1995. p 1424-1430.
22. King RWP, Tremblay BS, Strohbehn JW. The electromagnetic field of an insulated antenna in a conducting or dielectric medium. *IEEE Trans Microw Theory* 1983;31:574-583.
23. Atalar E. *Safe coaxial cables for MRI*. Chicago: Radiology; 1998. p 431-432.
24. Felmlee JP, Hakanson D, Zink FE, Perkins WJ. Real time evaluation of EKG electrode heating during MRI at 1.5T. In: *Proc ISMRM, 3rd Scientific Meeting and Exhibition*, Nice, 1995. p 1226.
25. Fitzsimmons JR. The design of RF systems for patient safety. *Ann NY Acad Sci* 1992;649:313-321.

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